# METHOD AND APPARATUS FOR CORRECTING A VOLUMETRIC SCAN OF AN OBJECT MOVING AT AN UNEVEN PERIOD

## BACKGROUND OF THE INVENTION

[0001] This invention relates generally to diagnostic ultrasound systems. In particular, the present invention relates to method and apparatus for obtaining a volumetric scan of a periodically moving object within a body.

[0002] A current challenge exists to perform a multidimensional ultrasound scan of a quickly, and more or less rhythmically, moving object within a body, such as a fetal heart. Currently, volume probes having a conventional one-dimensional (1D) array, which is mechanically moved in the elevation direction, as well as electronically steered 2D arrays, may be used for the acquisition. This technique makes it possible to acquire pyramid-shaped volume data sets. To image the fetal heart, high frame rate acquisitions are necessary, no matter whether 2D or 3D data sets are being acquired. For acquiring 3D data sets in real-time, one limitation is the constant speed of sound at 1540 m/s; this limits the amount of data to be acquired per second and thus these acquisitions are a tradeoff between frame rate and image quality. To acquire and achieve high frame rates, the line density has to be decreased, which significantly impairs lateral and elevation resolution.

[0003] One approach is to perform ECG-triggered volumetric acquisitions as described in US 5,159,931 to Pini, Nov. 3, 1992, which is hereby incorporated by reference in its entirety, which works well when imaging the adult heart, but is generally not available for the fetal heart due to absence of an appropriate fetal ECG signal. Alternatively, one may acquire data of several heart cycles at several fixed positions which are recorded with a position sensor, and obtain cardiac motion information via Fourier transform methods as described in Nelson et al, "Three Dimensional Echocardiographic Evaluation of Fetal Heart Anatomy and Function: Acquisition, Analysis, and Display", J Ultrasound Med 15:1-9, 1996, which

is hereby incorporated by reference in its entirety. Another problem is experienced when imaging a fetal heart during early pregnancy with 2D fetal echocardiography, when the relationship between fetus and amniotic fluid allows a lot of movement. If the fetus is very active, it may be time consuming or impossible at the time of the scheduled exam to acquire sufficient cardiac data.

[0004] Thus, a system and method are desired to obtain multidimensional data sets of a quickly moving object within a body that addresses the problems noted above and others previously experienced.

BRIEF DESCRIPTION OF THE INVENTION

[0005] In one embodiment, a method of processing a volumetric scan of a periodically moving object comprises performing a volumetric scan of a periodically moving object. A time interval of a periodic movement of the periodically moving object is identified within the volumetric scan. The volumetric scan is divided into volume subsets based on the time interval. The periodic movement is analyzed to determine a correction between adjacent volume subsets, and the time interval is adjusted based on the correction.

[0006] In one embodiment, a method of acquiring a diagnostic image of a periodically moving object comprises acquiring a series of scan planes comprising a moving object. The moving object repeats a cycle of movement over time, and the series of scan planes are acquired over at least two movement cycles. The series of scan planes are divided into N adjacent subsets, each having a first time interval. At least one common point of interest is identified within each of the series of scan planes, and intensity values of the at least one common point of interest are compared between the adjacent subsets. An adjusted time interval is calculated for at least one of the adjacent subsets based on the intensity values.

[0007] In one embodiment, an apparatus for acquiring a volumetric scan of a periodically moving object comprises a transducer. The transducer comprises an array of elements for transmitting and receiving ultrasound signals to

and from an area of interest comprising a periodically moving object. A transmitter drives the array of elements to scan the periodically moving object once in a single direction. A receiver receives the ultrasound signals which comprise a series of adjacent scan planes. A processor processes the series of adjacent scan planes. The processor identifies a time interval based on the periodically moving object and divides the series of adjacent scan planes into volume subsets based on the time interval. The processor also compares adjacent volume subsets and calculates adjusted time intervals for at least one of the adjacent volume subsets.

BRIEF DESCRIPTION OF THE DRAWINGS

[0008] FIG. 1 illustrates a block diagram of an ultrasound system formed in accordance with an embodiment of the present invention.

[0009] FIG. 2 illustrates an ultrasound system formed in accordance with an embodiment of the present invention.

[0010] FIG. 3 illustrates a series of projecting scan planes acquired by the ultrasound system of FIGS. 1 and/or 2 in accordance with an embodiment of the present invention.

[0011] FIG. 4 illustrates a series of adjacent scan planes acquired in accordance with an embodiment of the present invention.

[0012] FIG. 5 illustrates a 2D slice of the adjacent scan planes of FIG. 4 in accordance with an embodiment of the present invention.

[0013] FIG. 6 illustrates a plot of intensity over time and space for an x,y point in accordance with an embodiment of the present invention.

[0014] FIG. 7 illustrates an autocorrelation of an intensity line of an x,y point in accordance with an embodiment of the present invention.

[0015] FIG. 8 illustrates an autocorrelation of an intensity line of an x,y point in accordance with an embodiment of the present invention.

[0016]   FIG. 9 illustrates a single 2D image in accordance with an embodiment of the present invention.

[0017]   FIG. 10 illustrates a summed autocorrelation in accordance with an embodiment of the present invention.

[0018]   FIG. 11 illustrates a series of adjacent scan planes acquired in accordance with an embodiment of the present invention.

[0019]   FIG. 12 illustrates a series of volumes corresponding to the subsets of scan planes of FIG. 11 in accordance with an embodiment of the present invention.

[0020]   FIG. 13 illustrates a series of scan planes of a fetal heart in accordance with an embodiment of the present invention.

[0021]   FIG. 14 illustrates a volume in accordance with an embodiment of the present invention.

[0022]   FIG. 15 illustrates a plot of intensity over time and space for the line through the volume subset of FIG. 14 in accordance with an embodiment of the present invention.

[0023]   FIG. 16 illustrates a plot of intensity over time and space for the line through the volume subset, which is the volume subset adjacent to volume subset, of FIG. 14, in accordance with an embodiment of the present invention.

[0024]   FIG. 17 illustrates a cross correlation of intensity lines in accordance with an embodiment of the present invention.

[0025]   FIG. 18 illustrates a plot of intensity over time and space for the line through the volume subset of FIG. 14 in accordance with an embodiment of the present invention.

[0026]   FIG. 19 illustrates a plot of intensity over time and space for the line through the volume subset of FIG. 14 in accordance with an embodiment of the present invention.

[0027]   FIG. 20 illustrates a plot of intensity over time and space for the line through the volume subset of FIG. 1 in accordance with an embodiment of the present invention.

[0028]   FIG. 21 illustrates a plot of intensity over time and space for the line through the volume subset of FIG. 14 in accordance with an embodiment of the present invention.

[0029]   FIG. 22 illustrates a summed cross correlation in accordance with an embodiment of the present invention.

[0030]   FIG. 23 illustrates adjusted volume subsets within the volume in accordance with an embodiment of the present invention.

DETAILED DESCRIPTION OF THE INVENTION

[0031]   FIG. 1 illustrates a block diagram of an ultrasound system 100 formed in accordance with an embodiment of the present invention.  The ultrasound system 100 includes a transmitter 102 which drives transducers 104 within a probe 106 to emit pulsed ultrasonic signals into a body.  A variety of geometries may be used.  The ultrasonic signals are back-scattered from structures in the body, like blood cells or muscular tissue, to produce echoes which return to the transducers 104.  The echoes are received by a receiver 108.  The received echoes are passed through a beamformer 110, which performs beamforming and outputs an RF signal.  The RF signal then passes through an RF processor 112.  Alternatively, the RF processor 112 may include a complex demodulator (not shown) that demodulates the RF signal to form IQ data pairs representative of the echo signals.  The RF or IQ signal data may then be routed directly to RF/IQ buffer 114 for temporary storage.

[0032] The ultrasound system 100 also includes a signal processor 116 to process the acquired ultrasound information (i.e., RF signal data or IQ data pairs) and prepare frames of ultrasound information for display on display system 118. The signal processor 116 is adapted to perform one or more processing operations according to a plurality of selectable ultrasound modalities on the acquired ultrasound information. Therefore, the signal processor 116 may be used to perform the functions of a STIC analyzer and converter 42 and a volume display processor 46, which are described below. Acquired ultrasound information may be processed in real-time during a scanning session as the echo signals are received. Additionally or alternatively, the ultrasound information may be stored temporarily in RF/IQ buffer 114 during a scanning session and processed in less than real-time in a live or off-line operation. An image buffer 122 is included for storing processed frames of acquired ultrasound information that are not scheduled to be displayed immediately. The image buffer 122 may comprise any known data storage medium.

[0033] FIG. 2 illustrates an ultrasound system formed in accordance with one embodiment of the present invention. The system includes a probe 10 connected to a transmitter 12 and a receiver 14. The probe 10 transmits ultrasonic pulses and receives echoes from structures inside of a scanned ultrasound volume 16. Memory 20 stores ultrasound data from the receiver 14 derived from the scanned ultrasound volume 16. The volume 16 may be obtained by various techniques (e.g., 3D scanning, real-time 3D imaging, volume scanning, 2D scanning with transducers having positioning sensors, freehand scanning using a Voxel correlation technique, 2D or matrix array transducers and the like).

[0034] The transducer 10 is moved, such as along a linear or arcuate path, while scanning a region of interest (ROI). At each linear or arcuate position, the transducer 10 obtains scan planes 18. The scan planes 18 are stored in the memory 20, and then passed to a spatial and temporal image correlation (STIC) analyzer and converter 42. In some embodiments, the transducer 10 may obtain lines instead of the scan planes 18, and the memory 20 may store lines obtained by the transducer 10 rather than the scan planes 18. Data output by the STIC analyzer and converter 42 is

stored in volume memory 44 and is accessed by a volume display processor 46. The volume display processor 46 performs volume rendering and/or other image processing techniques upon the data. The output of the volume display processor 46 is passed to the video processor 50 and display 67.

[0035] The position of each echo signal sample (Voxel) is defined in terms of geometrical accuracy (i.e., the distance from one Voxel to the next), ultrasonic response, and derived values from the ultrasonic response. Suitable ultrasonic responses may include B-flow, gray scale values, color flow values, and angio or power Doppler information.

[0036] FIG. 3 illustrates a series of projecting scan planes acquired by the ultrasound system 100 of FIGS. 1 and/or 2. Each line 150-156 represents a scan plane into the page, and the elevation 164 of the scan is indicated. Although FIG. 3 illustrates the scan planes in a fan-shape, it should be understood that the fan-shape is not a limitation, and that other shapes, such as a rectangle with parallel scan planes, for example, may also be acquired. The following discussion will be based on acquiring data representing the fetal heart, but it should be understood that other periodically moving objects may be similarly scanned and processed, such as an adult heart, a heart valve, an artery, a vein and the like. Also, although the modality discussed is ultrasound, the image acquisition and processing techniques may be used with other modalities, such as CT, MRI, and the like.

[0037] Similarly, the information acquired by the ultrasound system 100 is not limited to B-mode information only, but may also contain information gathered from evaluating several lines from the same sample volume (e.g., color Doppler, power Doppler, tissue Doppler, B-flow, Coded Excitation, harmonic imaging, and the like). These data of different ultrasound modalities or scanning techniques may also be acquired simultaneously, and may be used either for analysis, display, or both.

[0038] The probe 10 is held in one position throughout the acquisition, and is positioned to acquire data representative of the item of interest,

such as the fetal heart. The transducers 104, or array of transducers 104, are electronically or mechanically focussed to direct ultrasound firings longitudinally to scan along adjacent scan planes, and external position sensing is not necessary.

[0039] A single, slow, acquisition sweep acquiring adjacent scan planes 18 may start, by way of example only, at border 158, and end at border 160. Other start and end points of the acquisition sweep may also be used. By way of example only, the acquisition sweep may have a sweep angle 162 of 20 degrees and a time period including several, or at least two, movement cycles of the fetal heart. Other sweep angles 162 may be used. The acquisition sweep may be accomplished by continuously moving the focus of the ultrasound firings or by changing the focus in small increments.

[0040] Alternatively, the acquisition sweep may have an acquisition time period covering multiple movement cycles, and the sweep angle 162 may be changed to reflect the type and/or size of anatomy being scanned. An acquisition with a longer acquisition time will acquire more data and the spatial resolution will be better when compared to a scan acquired over a shorter acquisition time. An acquisition with a higher frame rate will result in better temporal resolution than a scan acquired with a lower frame rate. The transducers 104 are focussed to acquire the adjacent scan planes 18 very close to each other spatially.

[0041] Each scan plane represented by the lines 150-156 in FIG. 3 is a 2D image having a resolution in time. By way of example only, if the resolution in time is within a range of 50-150 Hz, 50-150 scan planes, respectively, may be acquired for every 2 degrees of sweep angle 162, for an acquisition time of 10 seconds and a sweep angle 162 of 20 degrees. The resolution in time is not limited by this example. For clarity, not all of the scan planes are illustrated.

[0042] FIG. 13 illustrates a series of scan planes 310 of a fetal heart. The beating of the fetal heart during the acquisition causes a rhythmical pattern of changes in the diameter of the heart. The scan planes are acquired in the combined

direction of space and time 312. Markers 314 and 316 are included to illustrate the fetal heart rate. Calculating the fetal heart rate is discussed below.

[0043] FIG. 4 illustrates a series of adjacent scan planes 170, such as the adjacent scan planes acquired in FIG. 3. The scan planes are illustrated over combined space and time 176 with axial 172 and azimuth 174 directions as indicated. Therefore, the elevation is orthogonal to the scan planes 170. A 2D image 178 having data indicative of one scan plane of the fetal heart is represented at the end of the adjacent scan planes 170. Referring again to FIG. 13, 2D image 318 also comprises data of one scan plane of the fetal heart.

[0044] FIG. 5 illustrates a 2D slice 190 of the adjacent scan planes 170 of FIG. 4. An image correlation technique is used to extract the movement of the fetal heart from the scan plane data.

[0045] Returning to FIG. 4, the signal processor 116 chooses an x,y coordinate, point 182, from image acquisition system coordinates. By way of example only, the coordinates may be polar, rectangular, and the like, and are not limited to a specific acquisition geometry. The point 182 is identified on each scan plane. In FIG. 5, line 192 is illustrated. Line 192 corresponds to points 182, and therefore runs through the series of adjacent scan planes 170 at the same x,y coordinate.

[0046] FIG. 6 illustrates a plot 200 of intensity 194 over time and space 196 for the point 182. Therefore, an intensity value is identified for the point 182 on each image frame and displayed over time as intensity line 198. The periodic movement of the fetal heart introduces periodic intensity variation (as illustrated by the rhythmical pattern of the intensity line 198). The variations in intensity are analyzed by the STIC analyzer and converter 42 in order to obtain movement information. It should be understood that the plot 200 is a representation only, and that other methods, such as storing the identified intensity values in memory 20, volume memory 44, or image buffer 122 may be used.

[0047]    FIG. 7 illustrates an autocorrelation 184 of intensity line 198. The autocorrelation 184 may be calculated by the STIC analyzer and converter 42  by taking the autocorrelation function of the intensity line 198, such as by using the following equation:

$$A(y) = \int s(x) \cdot \overline{s(x-y)} dx$$

wherein A(y) is the autocorrelation function of the signal s, x is the integration variable in the spatio-temporal domain, y is the lag of the autocorrelation function, and s is the intensity line 198.

[0048]    By calculating the autocorrelation 184, a peak 202 at a zero-position is identified.  The peak 202 is the highest peak, or the peak with the most energy.  The STIC analyzer and converter 42 then identifies a first significant local maximum 204, which is the peak with the next highest energy.  The STIC analyzer and converter 42 calculates a time interval 206 between the peak 202 and the first significant local maximum 204.  The time interval 206 identifies the period of the heart cycle.  Once the time interval 206 is known, the STIC analyzer and converter 42 determines how many adjacent scan planes were acquired within the time interval 206.

[0049]    Alternatively, the heart cycle may be calculated using Fast Fourier Transform Analysis (FFT).  The STIC analyzer and converter 42  or signal processor 116 may identify the frequency of the movement as the location of the first significant local maximum 204 in a power spectrum of the intensity line 198.  In addition, Doppler methods may be used to determine the velocity of tissue movement to identify specific motion states (e.g., Systole, diastole, etc) of the object.

[0050]    FIG. 8 illustrates an autocorrelation 210 of intensity line 208 (FIG. 5).  The autocorrelation function was taken of the intensity line 208, which corresponds to a point 186 (FIG. 4) identified outside the heart.  Therefore, there is little or no periodic movement, and intensity line 208 may only have one significant peak 212. A time interval cannot be calculated.

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[0051] FIG. 9 illustrates a single 2D image 220 similar to the 2D image 178 of FIG. 4. For simplicity, it may be assumed that the heart or other moving object of interest is approximately in the center of the 2D image 220. Therefore, the signal processor defines a border 222 around an interior portion 232 of the 2D image 220. The border 222 may be a predefined number of pixels in from the top 224, bottom 226 and sides 228, 230 towards the interior portion 232 of the 2D image 220. The border 222 need not be symmetrical along all edges.

[0052] The STIC analyzer and converter 42 defines a number of points 234 (similar to points 182 and 186 of FIG. 4) within the interior portion 232. The number of points 234 may be defined by a pattern having a defined size and resolution. For example, the pattern may resemble a chess board in which every other point is chosen. Alternatively, every $4^{th}$ or $10^{th}$ point may be chosen. The pattern may change based upon user preference, or upon the anatomy being scanned. For example, a different template may be offered for a fetal heart, heart valve, artery, and the like. Optionally, the STIC analyzer and converter 42 may randomly select a predefined number of points 234 within the interior portion 232. The number of points 234 and/or the size and resolution of the pattern may vary depending upon factors such as processing speed, image resolution, and the like.

[0053] Returning to FIG. 5, a line (similar to lines 192 and 208) is defined for each point in the number of points 234 (FIG. 9). An intensity line 198 (FIG. 6), or intensity values, are defined for each of the lines corresponding to the number of points 234. The autocorrelation 184 is then taken of each intensity line 198.

[0054] It should be understood that although FIG. 9 represents a 2D image, the above method may be applied to data acquired using other acquisition modes, such as Doppler, B-flow, and the like.

[0055] FIG. 10 illustrates a summed autocorrelation 240. The STIC analyzer and converter 42 sums the autocorrelations taken for each point in the number of points 234. By summing the autocorrelations, noise 214 (FIG. 7) is

effectively removed from the summed autocorrelation 240. In addition, points located in areas not experiencing movement, such as above or below the heart (point 186) do not prevent the STIC analyzer and converter 42 from calculating the average time interval 246. Alternatively, a filter or windowing function may be used in addition to remove noise 214.

[0056] The STIC analyzer and converter 42 identifies a peak 242 at zero (having the highest intensity) and a first significant local maximum 244 as discussed previously. The time interval calculated between the zero peak 242 and the first significant local maximum 244 is the average time interval 246 of one movement cycle. The STIC analyzer and converter 42 now determines the number of scan planes occurring within the average time interval 246, or heart cycle.

[0057] FIG. 11 illustrates a series of adjacent scan planes 250 similar to the adjacent scan planes 170 of FIG. 4. Time intervals 252-258 separate the series of adjacent scan planes 250 into the number of slices determined by the STIC analyzer and converter 42. The location of time intervals 252-258 within the series of adjacent scan planes 250 need not correspond to the start and end of the heart cycle. Therefore, the STIC analyzer and converter 42 may choose any scan plane within the series of adjacent scan planes 250 to begin sectioning the data into time intervals 252-258. The STIC analyzer and converter 42 may move forward and backward through the series of adjacent scan planes 250 to section the data into the time intervals 252-258. By way of example only, the series of adjacent scan planes 250 comprise 4 scan planes within each time interval 252-258. Time interval 252 comprises scan planes 260-266, time interval 254 comprises scan planes 268-274, time interval 256 comprises scan planes 276-282, and time interval 258 comprises scan planes 284-290. When imaging a fetal heart or other anatomy as discussed previously, however, many more scan planes would be acquired within each time interval 252-258.

[0058] The STIC analyzer and converter 42 rearranges the order of the scan planes 260-290 and combines the scan planes acquired at the same phase, or point in time within the heart cycle, but from a different lateral position, into a volume. As illustrated in FIG. 11, scan planes A1 260, B1 268, C1 276, and D1 284

were acquired during the same phase within the heart cycle. Similarly, each of the following subsets of scan planes [scan planes A2 262, B2 270, C2 278, D2 286], [scan planes A3 264, B3 272, C3 280, D3 288], and [scan planes A4 266, B4 274, C4 282, D4 290] were acquired during the same phase within the heart cycle.

[0059] FIG. 12 illustrates a series of volumes 292 corresponding to the subsets of scan planes 260-290 of FIG. 11. While the volume 250 of FIG. 11 incorporates two purely spatial dimensions and one combined spatio-temporal dimension, each of the volumes 294-300 of FIG. 12 has purely 3 spatial dimensions and covers data from a single point in time (relative to the heart cycle).

[0060] The STIC analyzer and converter 42 combines each subset of scan planes 260-290 into one volume. Therefore, volume 1 294 comprises the image data of scan planes A1 260, B1 268, C1 276, and D1 284. Volume 2 296 comprises the image data of scan planes A2 262, B2 270, C2 278, and D2 286. Volume 3 298 comprises the image data of scan planes A3 264, B3 272, C3 280, and D3 288. Volume 4 comprises the image data of scan planes A4 266, B4 274, C4 282, and D4 290. Each volume 294-300 comprises a snapshot of the fetal heart during one single beat.

[0061] The series of volumes 292 may be displayed in three orthogonal planes in a cycle, such as a cineloop, which allows a user to navigate through the volumes 294-300 and view individual volumes. When imaging a fetal heart, for example, approximately 40-60 volumes 294-300 may be created and displayed on display 67. Each of the 40-60 volumes 294-300 represents a fixed point within the heart cycle.

[0062] Alternatively, the data may be processed and displayed in other ways. For example, the volume display processor 46 may render the image data to show the inner 3D structure of the heart. For example, maximum intensity projection, minimum intensity projection, average projection, and the like may be calculated and displayed. Also, a single volume 294-300, or a portion or slice of a volume 294-300, may be selected for display. The selected portion may be rotated on

the display 67 or further processed separate from the remaining volume data. In addition, an anatomic M-mode image representing a single selected point within the volumes may be displayed over time. The heart rate and/or other data may also be displayed.

[0063] Now that the series of volumes 292 has been created, it may be stored in a memory, such as ultrasound data memory 20, image buffer 122, a hard drive, a floppy, CD, or DVD drive, or on a server on a network. The series of volumes 292 and/or the unprocessed volumetric data may also be transferred via a network or one of the aforementioned portable discs to be further processed and reviewed at a different location after the patient has left the examination. Having the data available to be reviewed and processed later is advantageous, especially during early pregnancy, when the relationship between fetus and amniotic fluid allows a lot of movement. Once the data is acquired, fetal movement is no longer an issue, as previously discussed in relation to 2D fetal echocardiography.

[0064] The aforementioned method and apparatus may be used to calculate a fetal heart rate or a period of a repeating cycle of movement over time for an object of interest. It is possible that the heart rate may vary across the volume of data, or that the repeating cycle of movement has varying intervals. Therefore, the following method and apparatus may be used to process a volume of data experiencing variations within the cycle of movement.

[0065] FIG. 14 illustrates a series of adjacent scan planes, or a volume 350. The signal processor 116 divides the volume 350 into volume subsets 352, 354, 356, and 358 with dividing lines 370, 372, and 374, respectively. Dividing line 376 may separate volume subset 358 and the next adjacent volume subset (not shown), or may illustrate an end to the volume 350. The volume subsets 352, 354, 356, and 358 have time intervals 360, 362, 364, and 366, respectively, which are equal, and may be calculated as discussed previously (FIGS. 9 and 10). Therefore, time intervals 360-366 represent the average motion period for volume 350, and each volume subset 352-358 includes the same number of adjacent scan planes.

[0066] Similar to point 182 of FIG. 4, the signal processor 116 defines a point (not shown) having an x,y coordinate. The point is identified on each adjacent scan plane. The signal processor 116 defines line 368, which corresponds to the identified point and runs through each scan plane of the volume 350 at the same x,y coordinate. Alternatively, the signal processor 116 may utilize a previously defined point, such as point 182 and 186.

[0067] FIG. 15 illustrates a plot 380 of intensity 382 over time and space 384 for the line 368 through the volume subset 352 of FIG. 14. FIG. 16 illustrates a plot 390 of intensity 392 over time and space 394 for the line 368 through the volume subset 354, which is adjacent to volume subset 352. Therefore, the signal processor 116 identifies an intensity value for the line 368 through each image frame within each of the adjacent volume subsets 352 and 354, and plots intensity lines 386 and 396, respectively. As stated previously with FIG. 6, the plots 380 and 390 are a representation only and other methods may be used.

[0068] FIG. 17 illustrates the cross correlation 400 of intensity lines 386 and 396. The cross correlation 400 may be calculated using the equation:

$$C(y) = \int s_1(x) \cdot \overline{s_2(x-y)} dx$$

where $s_1$ is the intensity line 386, $s_2$ is the second intensity line 396, and y is the lag of the cross correlation function. The signal processor 116 or STIC analyzer and converter 42 calculates the cross correlation 400 and then identifies the maximum 402. The maximum 402 characterizes the shift in relation to the zero point 404 of the equation at which the two intensity lines 386 and 396 match best.

[0069] The signal processor 116 (or STIC analyzer and converter 42) then calculates a distance between the zero point 404 and the maximum 402. The distance is a delta, or correction 406, between the heart cycle lengths within volume subsets 352 and 354. The correction 406 may be represented by acquisition time, or by a number of adjacent scan planes. It should be noted that if the volume subsets 352-358 illustrate heart cycles, the heart cycles comprising volume subsets 352 and 354 were not equal in time.

[0070]　The dividing line 370 between the volume subsets 352 and 354, and all successive dividing lines 372–374 are moved according to the correction 406. For example, if the maximum 402 is at a positive value, the dividing lines 370-374 are moved backwards or left (with respect to FIG. 14), and if the maximum 402 is at a negative value, the dividing lines 370-374 are moved forwards or right. Therefore, the dividing lines 370-374 are moved in the opposite direction with respect to the correction 406.

[0071]　Therefore, if the correction 406 is necessary, the positions of the dividing lines 370-376 within the volume 350 are disregarded, and adjusted dividing lines 462-468 now separate volume subsets 352-358. Volume subsets 352-358 now have adjusted time intervals 410, 411, 412 and 413, respectively. By way of example only, the adjusted time interval 410 is not equal to, and is represented as larger than, time interval 360 in FIG. 14. The adjusted time intervals 411-413 of the volume subsets 354-358 are each equal to their time intervals 362-366, respectively, and thus are equal to each other. In other words, the average motion period has been maintained for volume subsets 352-358.

[0072]　It should be noted that if the heart cycles comprising volume subsets 352 and 354 are equal in time, the maximum 402 and zero point 404 would be identified as the same point. Therefore, no correction 406 is required, and dividing line 370 is not adjusted.

[0073]　The signal processor 116 may adjust the volume subsets 352-358 to create adjusted volume subsets 414, 416, 418, and 419, respectively. The boundaries, such as the adjusted time intervals 410-413, identifying the adjusted volume subsets 414-419 are stored, for example, in slice memory 44 or image buffer 122. Optionally, one or more of the adjusted volume subsets 414-419 may also be stored.

[0074]　The signal processor 116 then plots the intensity 382 over time and space for the line 368 through the adjacent adjusted volume subsets 416 and 418 (FIGS. 16 and 17). The cross correlation 400 is calculated and the maximum 402

and zero point 404 are identified (FIG. 17). If the maximum 402 and zero point 404 are not identified as the same point, the correction 406 is calculated. The adjusted dividing line 464, and the successive (already adjusted) dividing lines 466-468 are moved accordingly. The adjusted dividing line 462, which separates adjusted volumes 414 and 416 which were previously compared to one another, is not moved. Adjusted volume subsets 416-419 and adjusted time intervals 464-468 may be created and stored, as previously discussed. For example, the previously adjusted volume subset 416 may be replaced with a new adjusted volume subset 416 based on the correction 406. The aforementioned method is repeated for each pair of adjacent volume subsets of the volume 350, until all of the volume subsets have been adjusted, if necessary.

[0075] To eliminate the negative effects of noise, and lines 368 which may be defined outside the heart, such as line 208 of FIG. 5, the signal processor 116 defines multiple points and lines. Returning to FIG. 9, the signal processor 116 may define the border 222 around the interior portion 232 of the 2D image 220. The number of points 234 are defined within the interior portion 232. Once again, the points 234 may be defined randomly or according to a pattern, with the type, size, and resolution of the pattern, or the number of points 234, possibly varying depending upon processing speed, image resolution, type of anatomy, and the like. Alternatively, the number of points 234 and the corresponding lines previously defined may be used. Also, the intensity values previously defined for each of the lines may be used.

[0076] Returning to FIG. 14, a line, such as lines 368, 378, and 388, is defined for each of the points 234. As previously discussed, FIGS. 15 and 16 illustrate intensity lines 386 and 396 for line 368 through volume subsets 352 and 354.

[0077] FIG. 18 illustrates a plot 420 of intensity 382 over time and space 384 for the line 378 through the volume subset 352 of FIG. 14. FIG. 19 illustrates a plot 424 of intensity 392 over time and space 394 for the line 378 through the volume subset 354 of FIG. 14. The signal processor 116 identifies intensity values

for line 378 through the volume subsets 352 and 354, and plots corresponding intensity lines 422 and 426.

[0078]    FIG. 20 illustrates a plot 428 of intensity 382 over time and space 384 for the line 388 through the volume subset 352 of FIG. 14.   FIG. 21 illustrates a plot 432 of intensity 392 over time and space 394 for the line 388 through the volume subset 354 of FIG. 14.  The signal processor 116 identifies intensity values for line 388 through the volume subsets 352 and 354 and plots corresponding intensity lines 430 and 434.  The signal processor 116 then calculates the cross correlation 400 (FIG. 17) for each pair of intensity lines (i.e. intensity lines 386 and 396 for line 368, intensity lines 422 and 426 for line 378, and so on).

[0079]    FIG. 22 illustrates a summed cross correlation 436.   The signal processor 116 sums the cross correlations 400 of all lines 368, 378, 388 within the volume subsets 352 and 354 which have previously been calculated.  The signal processor 116 identifies a maximum 438 of the summed cross correlation 436.  The maximum 438 is the average maximum for the cross correlations 400 calculated for all lines 368, 378, 388.  The signal processor 116 also identifies the zero point 440 for the summed cross correlation 436, which is defined as previously discussed with reference to FIG. 17.

[0080]    The signal processor 116 then calculates a distance between the zero point 440 and the maximum 438.  The distance is a delta, or correction 442, between the volume subsets 352 and 354.   The correction 442 may be represented by acquisition time, or by the number of adjacent scan planes, as previously discussed. The dividing lines 370-376 between the volume subsets 352-358 are moved according to the correction 442, resulting in the adjusted dividing lines 462-468 as previously discussed.  The locations of the dividing lines 464-468, which identify the adjusted volume subsets 414-419 are stored as previously discussed.  In addition, the one or more of the adjusted volume subsets 414-419 may also be stored.

[0081]    The signal processor 116 then plots the intensity 382 over time and space for the multiple lines 368, 378, 388 through the adjusted volume

subsets 416 and 418 (FIGS. 16 and 17). The cross correlations 400 are calculated and summed to create the summed cross correlation 436. The maximum 438 and zero point 440 are determined, and the correction 442 is calculated. The adjusted dividing lines 464-468 are adjusted if necessary, and the adjusted dividing lines 464-468 are calculated and stored. As discussed previously, the adjusted dividing line 462 is not moved, and the adjusted volume subset 414 is not changed. The aforementioned method is repeated for each pair of adjacent volume subsets of the volume 350, until all of the volume subsets have been adjusted, if necessary.

[0082]    FIG. 23 illustrates adjusted volume subsets 414 and 416 within the volume 350. The adjusted volume subset 414 has adjusted time interval 410 and comprises five scan planes 444-452. The adjusted volume subset 416 has adjusted time interval 411 and comprises four scan planes 454-460. It should be understood that a scan of the fetal heart or other anatomy would comprise many more scan planes within each adjusted time interval 410 and 411. However, the number of scan planes 444-460 in FIG. 23 have been limited for simplicity.

[0083]    The order of the scan planes 444-460 may now be rearranged into subsets of scan planes acquired at the same phase, or point in time, but from a different lateral position within the heart cycle. As the adjusted time intervals 410-413 of the volume subsets 414-419 are not all equal to one another, the signal processor 116 or STIC analyzer and converter 42 may interpolate between two adjacent scan planes to generate scan data from the right phase. The type of interpolation is not limited, and may be one of nearest neighbor interpolation, linear interpolation polynomial spline interpolation, and the like. The STIC analyzer and converter 42 receives each subset of scan planes and combines them into a single volume, such as the series of volumes 292 in FIG. 12.

[0084]    While the invention has been described in terms of various specific embodiments, those skilled in the art will recognize that the invention can be practiced with modification within the spirit and scope of the claims.